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Abstract: Motivated by a long term goal to develop a freehand 3-D ultrasound system which will be set up as a basic module in a neurosurgical environment we combined an existing tool which is dedicated to acquisition and visualization of 3-D Ultrasound, the so-called *StradX* with an optical tracker which is the most appropriate position tracker for such an environment. This report aims to transfer the experience that has been achieved during the construction of the freehand 3-D ultrasound set up and pinpoint crucial aspects which have been emerged during the period of experimenting in a clinical environment with cases which can be met in a daily routine. Our experimental results concern the examination of the transfontanelle of the brain and the carotid.

Key-words: 3-D freehand Ultrasound, acquisition, visualization, neuronavigation

(Résumé : tsvp)

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Towards Free-Hand 3-D Ultrasound

Résumé : Un des enjeux majeurs de la chirurgie assistée par l'image est de pouvoir réaliser des acquisitions d'images 3D en cours d'opération. Le travail présenté ici se place dans cette perspective et donne des indications pour réaliser l'implantation d'un système d'acquisition main libre en échographie 3D dédié à la reconstruction et à la visualisation d'images ultrasonnores 3D dans un contexte pouvant être utilisé en salle de chirurgie. Le système de visualisation testé est le système *StradX*. Pour ce qui concerne la sonde échographique, un traceur optique de position, associé à une caméra de neuronavigation, a été développé et testé. Ce rapport technique a pour objectif de transmettre notre expérience acquise au cours d'expérimentations en conditions cliniques afin de pointé du doigt certains aspects cruciaux mis en évidence au cours de ces expériences. Ces résultats expérimentaux concernent à la fois l'observation du cerveau d'un nourrisson par des images échographiques transfontanellaires ainsi que des acquisitions échographiques 3D de la carotide.

Mots-clé : Echographie 3D en mode main libre, acquisition d'images, visualisation, neuronavigation

Contents

1 Introduction

Ultrasound imaging has made tremendous progress in obtaining important diagnostic information and aiding during an image-guided surgery. It posseses unique qualities including real-time imaging, physiologic measurement, use of non-ionizing radiation without any known bioeffects, it is relatively non-invasive and ultrasound equipment is far less expensive to buy and maintain compared to MR and CT imaging systems. All these contribute to strengthen the critical role of ultrasound technology in medical imaging.

The development of 3-D ultrasound imaging is a way to address the disadvantages of conventional ultrasound imaging [3, 6]. One disadvantage of 2-D ultrasound imaging relates to the difficulty to develop a 3-D impression of the patient anatomy only by manipulating the ultrasound transducer. The typical approach to overcome this problem is to scan repeatedly through the region-of-interest and clarify the exact spatial relationships. Under some conditions, this process can be time-consuming and tedious, particularly with curved structures, when there is a subtle lesion in an organ, in the case that we have distortion of the normal anatomy and when there are structures not commonly seen. Furthermore, it is difficult to localize the thin ultrasound image in the organ and reproduce a particular image location at a later time, making the conventional 2-D exam a poor imaging modality for quantitative prospective or follow-up studies. The goal of 3-D ultrasound imaging is to allow the physician to have a straightforward view of the 3-D anatomy without the need to rescan the patient, thus reducing the variability of the conventional technique. In this way, we set the basis for more accurate quantitative analysis.

Acquisition of 3-D ultrasound is crucial for two reasons. First, to avoid geometric distortions due to the acquisition of tomographic images in arbitrary orientations in the case of 3-D imaging, their relative position and angulation must be accurately known. Second, to avoid artifacts and distortions due to involuntary motion, the image acquisition must be performed rapidly. Until now, one can choose among three solutions, to acquire a series of 2-D Ultrasound images. These are the free-hand acquisition, the mechanical localizers and 3-D probes. In this work, we have chosen to use the free-hand acquisition because of the following reasons: (i) It permits to use standard, commercially available ultrasound machines, (ii) it can be accurate, (iii) it combines scans with arbitrary position and orientation, (iv) it permits to scan unlimited volume of the body as well as complex patient surfaces and (v) the required probe has relatively low cost demands. Although the free-hand acquisition permits the selection of optimal views for the operator, there are two basic constraints on the 3-D system. First, the exact relative position and angulation for each acquired image must be precisely known and second, the operator must ensure that there will be no gaps left when scanning the anatomy under investigation.

This report aims to transfer the experience that has been achieved during the creation of a freehand 3-D ultrasound set up and pinpoint crucial aspects which have been emerged during the period of experimenting in a clinical environment with cases which can be met in a daily routine. We believe that it is worthwhile to discuss this because it may contribute to the maturity of the advantageous and currently under an increasing demand 3-D freehand ultrasound imaging.

The organization of this report is as follows: Section 2 deals with the materials and methods which were considered for the construction of our experimental set-up of freehand 3-D Ultrasound. Section 3 is dedicated to present our experimentation in a real clinical environment. Then, we end up with our conclusions and aspects of Future research in Section 4.

2 Materials and Methods

2.1 Introduction

The fundamental building blocks for the creation of a set up dedicated to freehand 3-D ultrasound are: (i) a position sensing device; (ii) any commercially available ultrasound machine; (iii) a video frame grabber; (iv) a module dedicated to calibration of acquired images with respect to the real patient coordinates and (v) a reconstruction - visualization module. In our system, blocks (iii)-(v) have been substituted by the respective functionalities of StradX [8]. This is a freehand 3-D ultrasound system developed at the Department of Engineering of Cambridge University by A. Gee and R. Prager which aims to provide an alternative to voxel-based 3-D ultrasound [7]. It was motivated by the shortcomings in the construction of a voxel array, namely speed, ease of use and inability to provide a generic interpolation method which will satisfy both speed and quality in reconstruction. The basic principle is to discard the resampling stage altogether and work with only the raw B-scans and their associated positions. Our final goal is to produce voxel arrays, therefore although we keep an eye in StradX-based visualisation (any-plane slicing, panorama imaging) we focus on the results of the 3-D reconstruction which has been implemented via a simple nearest neighbour algorithm. This was originally written by Graham Treece [12] and is available along with the StradX package. The original version of StradX was made to be interfaced to an electromagnetic position tracker. Motivated by image-guided surgery applications in the long term, we have contributed to an add-on of StradX which can be interfaced to an optical tracker. Details on this aspect are explained in the next Section which is dedicated to the acquisition of ultrasound images.

2.2 Optical sensing system - Acquisition

Until now, free-hand acquisitions have been performed by considering the above constraints through four basic approaches on the construction of the positioning (tracking) system. Thus, we may select among positioners which are based on: (i) the acoustic ranging; (ii) a mechanical articulated arm; (iii) a magnetic field sensor and (iv) an optical tracker. Acoustic sensors receive signals which are emitted by ultrasonic emitters and determine location via time-of-flight. Mechanical sensors determine the position of a sensor endpoint based upon measurements of joint angles and information regarding the kinematics of the device. Magnetic sensors measure electrical currents induced in receiver coils when the receiver is moved within a magnetic field generated by the emitter. Last, optical sensors track the positions of one or more actively or passively illuminated markers and use geometric triangulation to determine the locations of these markers.

We have decided to use an optical tracking system because it provides the most reliable and accurate position sensor available for medical applications in the case that there is always a free line of sight between the sensor and the operating table [11]. Additionally, our choice has been guided by the fact that the optical tracker does not involve any magnetic field for determination of position data and consequently does not permit any deformation of these data in the presence of metallic structures which is unavoidable in a surgical environment [2].

The FlashPoint 3-D localizer includes three high resolution optical sensors that track the 3-D position in space up to 128 infrared LEDs [4]. These LEDs have been attached to the ultrasound probe. The principle of the localizer's operation is as follows: Each LED fires in sequence for several milliseconds. Since only one LED is illuminated at any given time, it is impossible for the systrem to make the mistake and consider one LED for another, even when they are in close proximity to each other. Some of the light from the LED passes through cylindrical lens in each sensor. In each case, the captured light is projected as a fine line onto the surface of a 5,000-element linear charge-coupled device (CCD) array mounted perpendicular to the lens. By detecting the location of the illuminated elements on the CCD, the sensor can measure the angle of incidence of the captured light raysrelative to the sensor. Given these three angular measurements and knowledge of the fixed location of each sensor relative to the others, the system can precisely measure the 3-D location of each LED in space. The location is reported as a relative distance and direction from the sensor array. The two outside sensors mounted on the optical sensor assembly are spaced roughly a meter apart. They are designed to provide a working volume of roughly a meter cubed, the center of which should be located about 1.5 meters away from the middle sensor. The system localizes the tip of the ultrasound probe by tracking the motion of the LEDs which have been mounted on the body of the device. Its accuracy is formally indicated in the range of 0.3 to 1.5 mm.

2.3 Calibration

Calibration is the stage that answers the question which is the correspondence between the coordinate system in the B-scans plane and the coordinate system of the physical space where the scanning takes place (position sensor readings). This answer has the form of 8 parameters which consist of the 6 existing degrees of freedom, three rotation (r, θ, ϕ) and three translation (x, y, z) plus the scaling parameters $(s_x u, s_y v)$.

Calibration is performed by scanning a phantom (an artificial object) of known geometric dimensions. There are different phantoms which can be used for calibration purposes. In [9] the interested reader may find a discussion about the subject.

The accuracy of calibration methods depends to some extent on the numerical methods employed. During our experiments, we have utilized a three stage optimization process which can be applied by using the StradX environment [9].

An additional aspect which is important for the calibration process is that special attention has to be paid in order to exercise all six degrees of freedom. Details of the whole process of calibration will be given at the description of the experimental work in Section 3

2.4 Visualization of 3-D ultrasound data

Three dimensional ultrasound data can be visualized in four different ways. These are : (a) any plane slicing, (b) panoramic imaging, (c) surface rendering and (d) volume rendering. While the first two methods do not require any 3-D reconstruction, (c) and (d) are depended on the generation of a 3-D representation from the acquired set of 2-D images. Using the first method (any plane slicing), we may display the ultrasound data on any plane selected by the user through the series of 2-D raw acquired data. In the second method, panoramic imaging or Extended Field of View (XFOV) imaging (also known as Siescape) [5] allows many B-scans to be aggregated obtaining a wide field of view.

In the case, that we are able to segment a series of 2-D B-scans in order to extract desired features, we can construct a 3-D surface model out of the boundary descriptions of the segmented regions. This model can be viewed via a surface rendering. Advantages of this way of viewing the ultrasound data include reduction of the amount of information which allows efficient rendering and production of 3-D images with good contrast between segmented structures. The only disadvantge appears to be its high dependence from the segmentation process which could be either of bad quality or highly time-consuming.

The last approach which is used to visualize ultrasound data is based on the construction of a 3-D voxel-based Cartesian volume. The volume can be built by placing each acquired 2-D image in its correct location in the volume. The gray scale values of voxels which are not sampled by the 2-D slices are calculated by interpolation between the appropriate images. The interpolation methods which have been mentioned in the literature that deal with reconstruction of 3-D Ultrasound data fall into four categories [10] : voxel nearest neighbour interpolation, pixel nearest neighbour interpolation, distance weighted interpolation and radial basis function interpolation. Reconstruction produces large data files and most methods which provide very good results are rather time-consuming leading to imaging that real-time is far from being a reality.

3 Experimental results

The whole experimental work took place at the University Hospital of Rennes (CHU) with the collaboration of the physicians Dr. P. Darnaud and C. Treguier. Our set-up for freehand Ultrasound 3-D consisted of the FlashPoint optical tracker system which had been interfaced to an SGI machine that permitted us to run our new version of StradX dedicated to acquire readings from the optical tracker. Tracking occured via 4 LEDs which had been mounted in fixed locations on the ultrasound probe shaft via a V-shaped support (see Figure 1). Using more than 2 LEDs enables us to capture 6 degrees of freedom, translation (x,y,z) and rotation (elevation, roll, azimuth). The video output of an HDI 5000 ultrasound machine was interfaced to the SGI machine which played the role of frame grabber via the respective capability of StradX. This set-up can be viewed in Figure 2.

During examination, our experimental work followed two main stages : the first stage was dedicated to calibration of the ultrasound probe and the second stage was the formal acquisition procedure. The completion of these stages permitted visualization of the data, examples of which are going to be presented in the sequel.

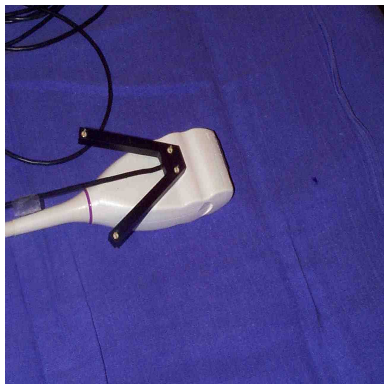


Figure 1: *The ultrasound probe with the attached LEDs assembly*

Figure 2: *The set-up for the experiments in a standard consultation room*

3.1 Calibration/Acquisition stage

As we have mentioned in Section 2.3, to perform calibration we need a phantom. For this purpose, we decided to use the *single wall* phantom [9] which is a simple water bath with planar surfaces. Advantages of this is its ease to find or construct and additionally scanning of the floor can produce clear and consistent lines in the B-scans. An example of an image which has been acquired using US to scan a single wall phantom can be seen in Figure 4(a). Satisfaction of the demand for clear lines and consequently for clearly defined planes facilitates automatic detection of the plane which results in a calibration process which is both rapid and easy to perform. Automatic line detection can be done via StradX and leads to an automatic calibration process. Figure 4(b) demonstrates the good performance of the line (plane) detection with StradX in the case of an acquisition with clear lines (Fig. 4(a)) using the single wall phantom.

Our experience in dealing with the calibration procedure as it is presented in StradX led us to the following conclusion : It is very important to exercise all the six degrees of freedom. Optical tracking systems are bound to the fact that need to have a free Line-of-Sight. The fact that we have constructed an assembly with all the LEDs attached on the same plane does not permit the availability of exercising all possible angles. A solution to this problem could be that a new construction is needed which may distribute LEDs in a spiral fashion around the probe. Such a construction has been reported by Beasley et al. [1]

A snapshot taken during the calibration procedure can be viewed in Figure 3.

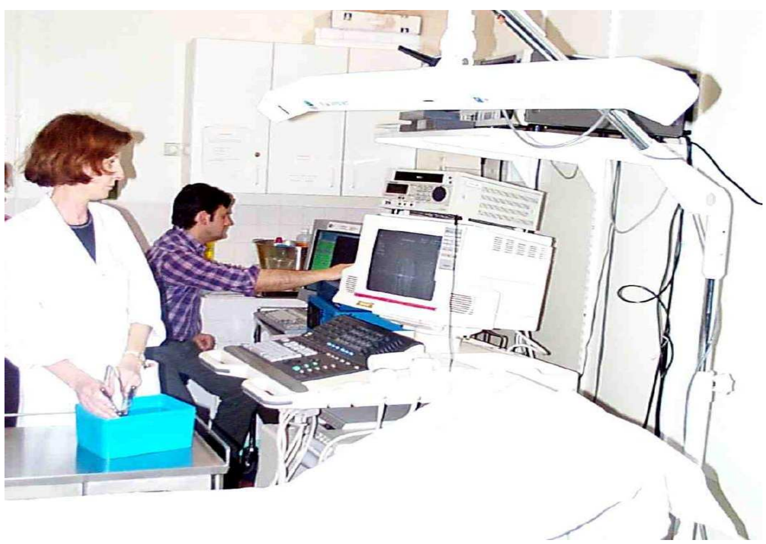
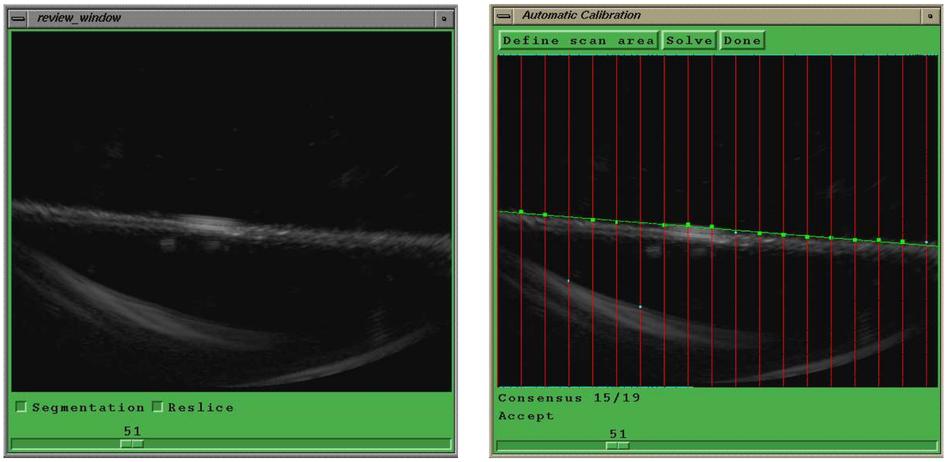


Figure 3: *Snapshot during the calibration procedure*



(a) (b)

Figure 4: *(a) US image of the thin wall phantom (b) Line fitting for the calibration*

Figure 5: *Acquiring US images during examination*

3.2 Visualization stage

In this report, we disseminate experimental results of two examinations. The first concerns the transfontanelle of the brain and the other concerns the carotid.

Concerning the first examination, we may see in Figure 6 one scan of the ensemble of the acquired B-scans with a clear definition of the ventricles, the sulci and the bones. In any such a scan we may draw a line which defines the plane over which the US data can be displayed. This is the so-called any-plane slicing. In Figure 7, we may observe a transversal section of the ventricles after the reslicing guided by the plane defined in Figure 6. In Figure 8 we present a mosaic of visualizations which can be realized in the StradX environment. Apart of the conventional B-scan view (bottom-left window) and its respective reslicing (top-right window), this mosaic also consists of a surface rendering of the ventricles (bottom-right window) and a number of wireframe outlines which show the position of the scan images in 3-D space (middle-top window).

Surface rendering was applied after a manual segmentation of the ventricle. This rendering can clearly indicate till which extent the ventricular system has been scanned. Evidently, the acquired scans were not sufficient to cover the whole surface of the ventricles. This becomes more clear when looking at the wireframe outlines of the scans (middle-top window). We may observe that all the outlines have a common intersection point. This shows that the physician has tried to acquire the images by only rotating the probe without applying any translation.

Reconstruction of this examination's scans to a volumetric representation produced the results which can be seen in Figure 9. Density of scan aquisition resulted in a good estimation of the interpolated values, thus having a volume with good resolution in the three dimensions. This can be seen by observing the three orthogonal views of the volume as they appear in Figure 9.

Volume rendering of the reconstructed data emerge the ventricles and the whiter matter tract as it can be seen in Figure 9(Top-left window).

The examination of the carotid was the source of Figure 10 which consists of an ensemble of visualizations with respect to the acquired scans. As before, we can view the scan in the form of any slicing visualization (bottom-right and bottom-left window), surface rendering (bottom-middle), outline of the scans(top-right) and a review window wherein the operator may select planes for subsequent any-plane slicing visualization. Observing the outlines of the scans we can see that there is a pure translation of the probe which appears to have rather large gaps. Although in this case we have a complete scanning of the structure under investigation (carotid), we may see from the any-plane slicing visualizations that we achieved very poor resolution in the z-axis. This has been caused by the produced gaps between the scans which may be originated from the patient's movement or the inexperienced probe movement by the physician. Although the resolution is not satisfactory enough for quantitative studies we can very easily recognize the well known bifucation of the carotid by a surface rendering (Figure 10 (bottom-middle)) and by a volume rendering 11, as well. For the sake of clarity, in Figure 11 we have outlined the carotid over the orthogonal and the rendered views.

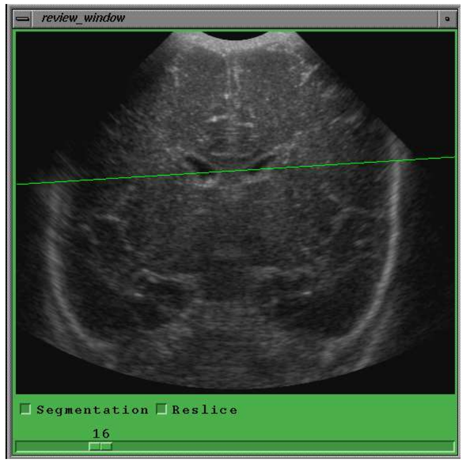


Figure 6: *An instant of reviewing the acquired images. The straight line indicates the plane over which a reslicing is applied*

Figure 7: *Visualisation via reslicing wrt to Figure 6*

Figure 8: *Ensemble of visualisations for the acquired scans*

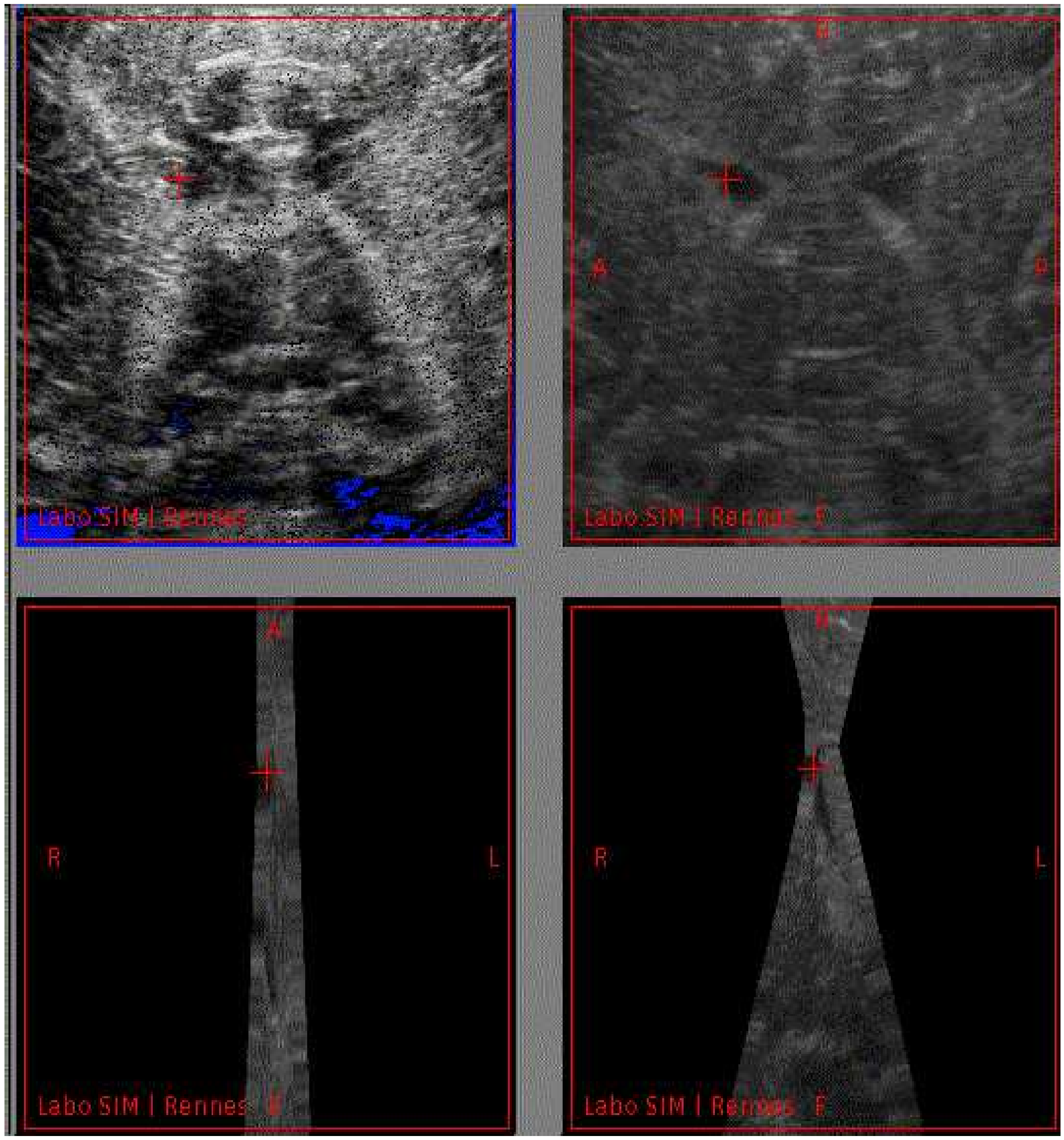


Figure 9: *Orthogonal views and rendering of US*

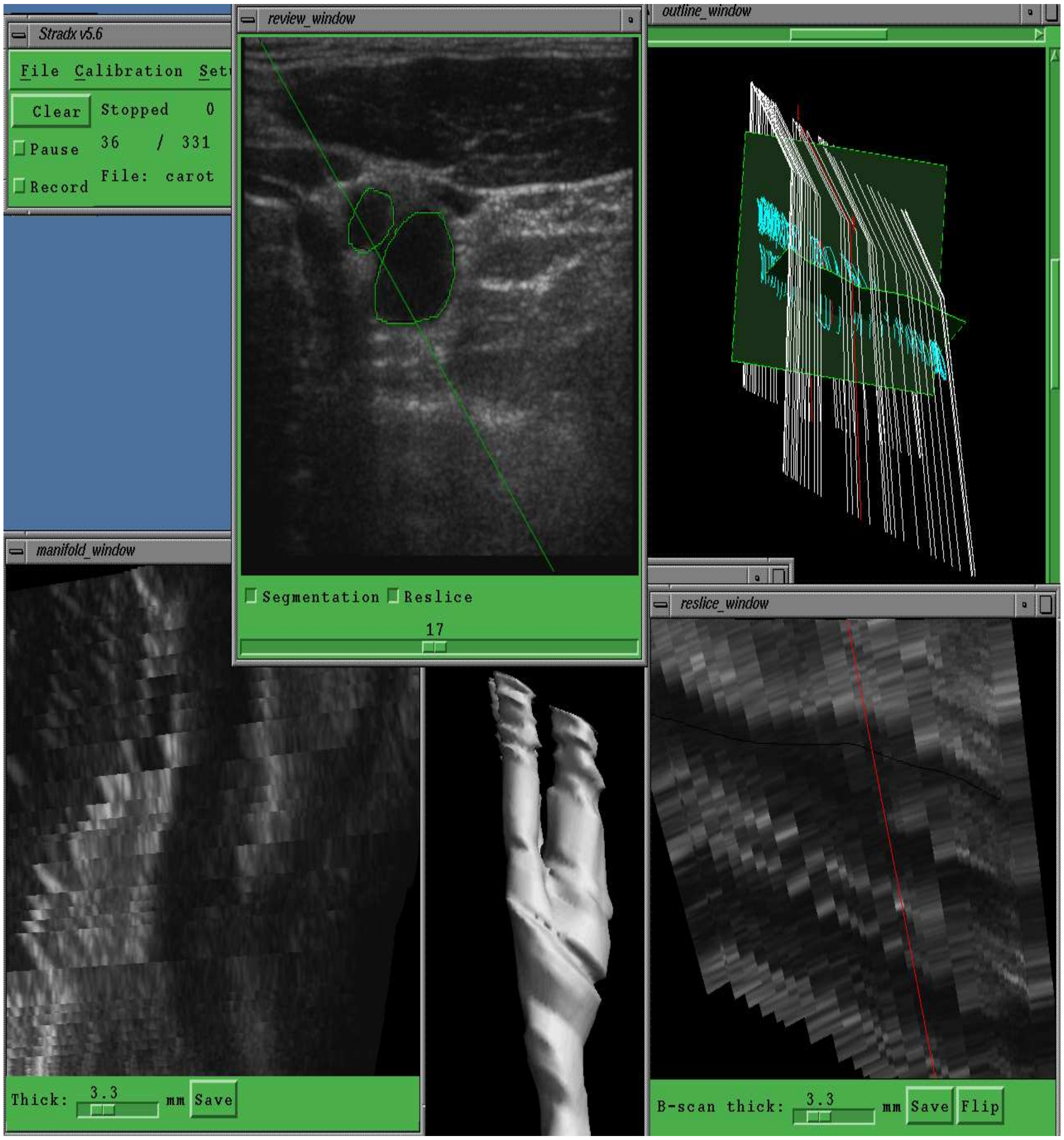


Figure 10: *Ensemble of visualisations for the acquired scans*

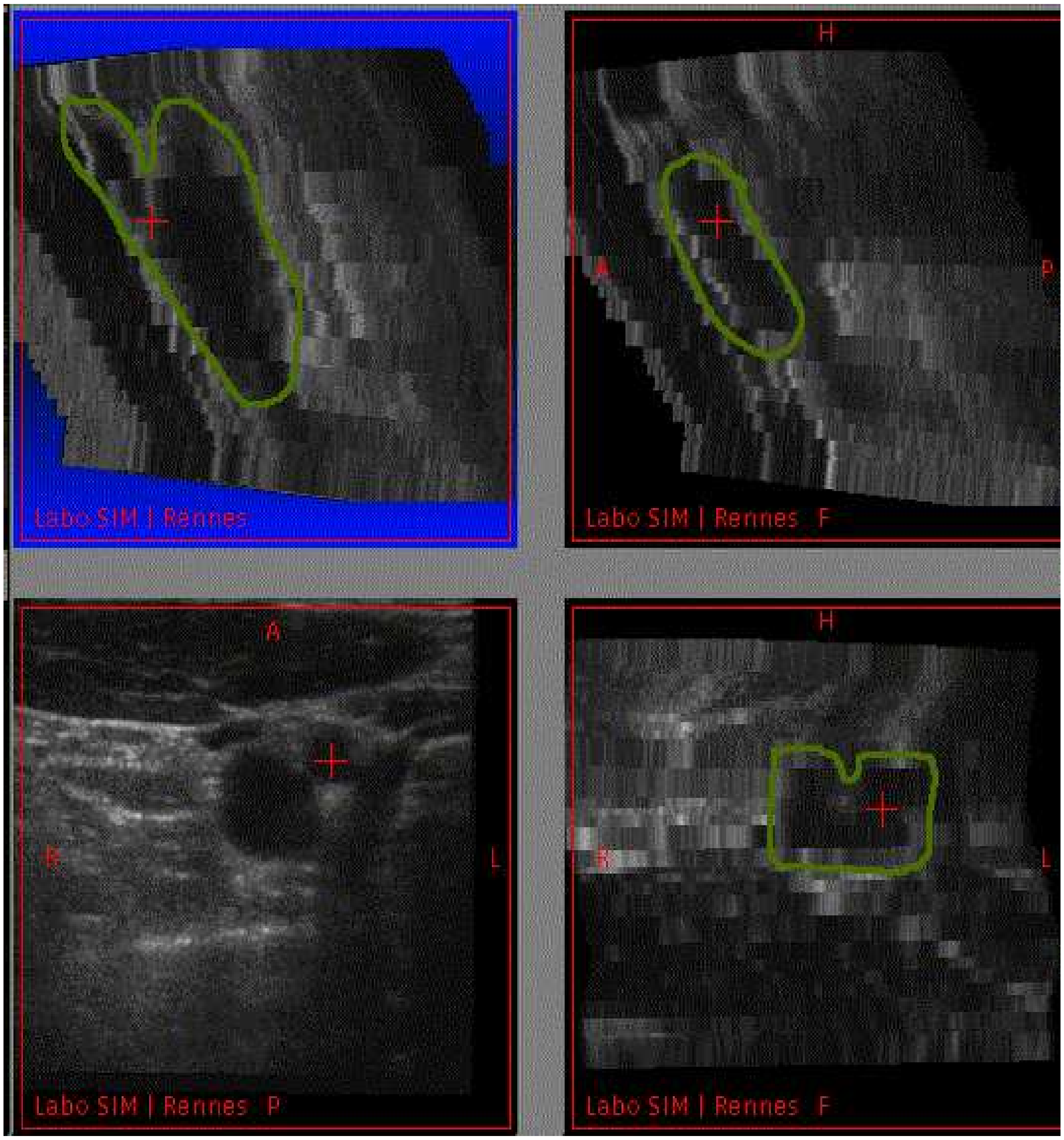


Figure 11: *Orthogonal views and rendering of US*

4 Conclusions

Motivated by a long term goal to develop a freehand 3-D ultrasound system which will be set up as a basic module in a neurosurgical environment we combined an existing tool which is dedicated to acquisition and visualization of 3-D Ultrasound, the so-called *StradX* with an optical tracker which is the most appropriate position tracker for such an environment. StradX offers several visualization capabilities, but none of these may serve voxel array constructions due to its basic underlying principle which does not favour volume reconstructions. Thus, although we experienced all the available visualisations of StradX, our interest was focused on the quality of the reconstructed volumes.

Our experimentation results do not lead in providing gold standards in the creation of freehand 3-D ultrasound systems. Rather, they provided us much experience by pinpointing existing pitfalls and advantages of the sequence calibration-acquisition-reconstruction-visualization.

The general conclusions which come out of this experimentation are :

- First of all, it becomes a task of high priority that the physicians must understand the basic principles of the 3-D ultrasound and avoid examining by using the conventional techniques of 2-D ultrasound. In practice, this can be done by applying a careful scanning considering not only rotations but also of translations. Also, they have to take care that they acquire regular and dense data sets. I point you to Section 3.2 in the example that we could not retrieve the ventricular structure due to lack of translations of the probe and the poor resolution in the z-axis due to non-dense data sets of the carotid.

- Calibration is depended on different requirements which have to be satisfied in order to reach an optimum result. These can be summarized in the following : (i) We have to exercize all the movements which can grasp the 6 degrees of freedom, (ii) we need to construct a LEDs assembly attached to the probe which will have a spiral distribution of the LEDs in order to capture unlimited angles, (iii) A careful probe movement may enable the single wall phantom be satisfactory enough.

- More robust Reconstruction techniques have to be applied. These have to compensate for the fundamental pitfalls of the scan series which are the gaps and the overlapping.

The above conclusions which came out after our experimentation with our freehand 3-D ultrasound will end up this report. Our hope is that this will activate fruitfull discussions and activities around the subject.

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